DOCTORAL THESIS

Dynamic foot and ankle characteristics in functionally relevant gait performance in those with and without a pathology

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by

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Abstract

The human ankle joint is hypothesized to be a primary controller of support, propulsion and steering during locomotion. A series of experiments were initiated to understand ankle plantarflexor muscle kinematics and kinetics in normal and pathological gait, and to define the specific locomotor demands of community ambulation. Additional experiments were then conducted to quantify the effects of walking speed on plantar pressures and centre of mass motion, to illuminate the role of the ankle in acceleration and deceleration during walking, and to examine how humans alter their kinematics and kinetics to turn. The results of these experiments provide support for the hypothesis that the ankle joint is important in a wide range of locomotor movements beyond walking straight ahead at constant speed. The ankle appears instrumental in adapting to different walking speeds, altering both the pressures on specific regions the plantar surface and the motion of the centre of mass across a range of speeds. The ankle also has subtle kinetic changes that appear to modulate acceleration and deceleration during single limb stance. For turning, the ankle plays a role during slowing into the turn and accelerating after the turn, but mediolateral shears appear to alter the trajectory of the body to negotiate a corner and the external hip rotators appear to rotate the trunk toward the new direction of travel. This work extends our understanding of the ankle in functionally relevant gait activities beyond simple straight-ahead walking at constant speed. The published papers included in this supporting statement have been cited by 180 different subsequent peer-reviewed publications, suggesting that this work has had some impact on the field.

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Introduction

The human foot-ankle system is a complex structure that is essential to optimal locomotor function. The ankle is the prime joint that controls the contact interface between the human body (the foot) and the locomotion surface (the ground), providing support[1-3], propulsion[4-7] and steering[8-11]. The neurological and structural components of the foot and ankle joint interact to control human locomotion and enable competent locomotor initiatives and responses to the environmental challenges of transport and mobility during the desired ambulatory movements of everyday life.

Early in my experiences of gait analysis I began to suspect that functional human ambulation required much more than walking straight ahead at self-selected speed. I visually observed that most human gait involved starting, turning, speed changes, and stopping: the gait of daily living. Without a method to investigate this or enough context to create an informed hypothesis, I began my research by investigating simple forms of walking before moving to inquiries of more complex walking movements that included different speeds, mechanisms to achieve a desired speed, and mechanisms to negotiate corners. This research campaign began with investigations of the influence of ankle pathology during walking, proceeded to a novel study of real-world ambulatory activities to demonstrate the ecological validity of the gait of daily living, and culminated with work on ankle function in more complex gait tasks.

The body of research presented here in this supporting statement highlight my contribution to the understanding of how the foot and ankle interact to control locomotion during the gait of daily living. Nine peer-reviewed papers published in a range of scientific journals form the foundation upon which this supporting statement is based.

Section 1. Normal plantarflexor function during walking

Orendurff MS, Segal AD, Aiona MD, Dorociak RD. Triceps surae force, length and velocity during walking. *Gait Posture.* 2005;21(2):157-163.

Yuen TJ, Orendurff MS. A comparison of gastrocnemius muscle-tendon unit length during gait using anatomic, cadaveric and MRI models. *Gait Posture.* 2006;23(1):112-117.

Initially I focused on defining the force, length and velocity characteristics of the plantarflexor muscle-tendon units during walking in normal adults. Several pathologies cause equinus, a condition marked by contractures of the plantarflexors, and surgical interventions designed to correct plantarflexor length also may influence muscle force production and shortening velocity during gait. Equinus causes a dysfunctional, inefficient gait and may warrant surgical correction, but the force-length-velocity characteristics of the *normal* human triceps surae during walking had not yet been defined. The force, length and velocity of a muscle-tendon unit are inexorably linked, and treating a plantarflexion contracture may have iatrogenic consequences.

A characterisation of the gastrocnemius and soleus force-length-velocity parameters during walking in normal individuals was needed for comparison to gastrocnemius and soleus force-length-velocity parameters after surgical intervention in pathological individuals. By having an objective force-length-velocity goal for triceps surae function during the gait cycle, more accurate decisions about the amount and type of surgical lengthening of the gastrocnemius and soleus could be made.

Therefore I initiated a project to describe the force, length and velocity characteristics of the gastrocnemius and soleus during walking for normal individuals so that a functional goal could be compared to the outcomes of interventions designed to improve plantarflexor kinematics and kinetics in pathological individuals. Briefly, the study utilized computerized gait analysis and muscle length change during gait was calculated using established methods[12], velocity was calculated using 3-point finite difference calculation and force was determined from sagittal ankle moment data. This paper was published in Gait and Posture[13] and has been cited by 10 peer-review publications (See Appendix A).

The citing publications cover the gamut of theoretical gait topics and have stirred considerable debate about the dynamical function of muscle tendon units during locomotion in pathology and unaffected individuals. These data have been focused on muscle parameters used in human gait modelling experiments and as normative data for comparison purposes in clinical studies. Several methods for calculating muscle length of the plantarflexors exist: one using models developed using cadaveric specimens[14], one using origin and insertion data collected on a larger number of specimens[15], and one using MRI to evaluate origin and insertion locations and wrapping of the muscle around other anatomical structures[12]. Each represents an improvement in accuracy based on improvements in sampling methods and technology. The debate as to the agreement between these methods was opened by my first publication, and a comparison was needed to illuminate differences between the methods. One of my students, Tracy Yuen, first-authored a paper with me comparing three different methods[12, 14, 15] of calculating the gastrocnemius and soleus lengths from sagittal ankle motion. This paper was nominated for best paper at the Gait and Clinical Movement Analysis Society

Conference, was published in Gait & Posture[16] and has been cited by six subsequent peer-reviewed publications (See Appendix A).

Section 2. Plantarflexor function in neuromuscular pathology

Orendurff MS, Aiona MD, Dorociak RD, Pierce RA. Length and force of the gastrocnemius and soleus during gait following tendo Achilles lengthenings in children with equinus. *Gait Posture.* 2002;15(2):130-135.

Armed with normative data and a comparison of various muscle length methods, my next investigation concentrated on neuromuscular pathology, specifically the effect of equinus contractures on ankle kinematics and kinetics during walking in children with spastic diplegia cerebral palsy, and in the risks and efficacy of surgical lengthening of the triceps surae. Therefore I initiated a project to determine the effects of surgical lengthening on the force, length and velocity characteristics of the plantarflexors during gait in children with cerebral palsy. This paper was published in Gait and Posture[17] and has been cited by 31 subsequent peer-reviewed publications (see Appendix A).

Using the same method as that for the normal individuals, my publication documents the change in the length characteristics, the force generating ability and the muscle-tendon unit velocity profile of the gastrocnemius and soleus during walking in children with spastic diplegia cerebral palsy before and after surgery. The conclusion of our study was that ankle power generation can be improved following judicious tendo Achilles surgical intervention to correct equinus, and that careful selection of appropriate candidates could increase muscle-tendon length without causing iatrogenic muscle weakness and crouch gait.

This finding is important because some are advocating only gastrocnemius aponeurotic lengthening to correct equinus in children with spastic cerebral palsy because they believe that all tendo Achilles lengthening will lead to iatrogenic crouch. They see no reason to even perform a Silvershold test (dorsiflexion with and without knee flexion to differentiate gastrocnemius from soleus tightness) since only one surgery should be performed. The advocates of this approach are very vocal and teach an annual course in gait analysis, and have a tremendous influence upon surgical decision-making in the clinical gait analysis community. Unfortunately, these clinician-scientists have not produced any peer-reviewed literature to support their position (although others have[18]), and some other published work has suggested that crouch can develop in the patient population even without any surgery whatsoever[19]. Recent published work has shown that recurrence of equinus is more likely than the development of iatrogenic crouch due to weakness or overlengthening,[20] at least for intramuscular (fascial) lengthenings of the gastrocnemius and soleus. Hopefully, my published work can add to the armamentarium of the orthopaedic surgeon for treatment of equinus in children with cerebral palsy. Careful consideration of the length, force producing capacity and contraction velocity may be valuable since these muscle-tendon unit characteristics are interconnected and surgery cannot alter length without altering force production and contraction velocity.

Section 3. Equinus and forefoot pressure in individuals with diabetes

Orendurff MS, Rohr ES, Sangeorzan BJ, Weaver K, Czerniecki JM. An equinus deformity of the ankle accounts for only a small amount of the increased forefoot plantar pressure in patients with diabetes. *J Bone Joint Surg Br.* 2006;88(1):65-68.

The triceps surae can also be affected by diabetes, which can alter the stiffness characteristics of the parallel elastic components (the extracellular membrane or cytoskeleton[21]) and the series elastic components (tendon[22-33]) due to the non-enzymatic glycolation of tissues. These advanced glycolytic end-products lead to equinus in individuals with diabetes, although the degree of toe-walking is not as severe as for individuals with cerebral palsy. Surgical lengthening of tendo Achilles or gastrocnemius aponeurosis has been suggested as a technique for reducing abnormal plantar pressure and therefore preventing ulceration beneath the metatarsal heads, a frequent cause of prolonged infection and eventually amputation in individuals with diabetes. It has been shown to be effective in reducing forefoot pressure by 27%, but this appeared to be temporary and it returned to more than 80 N/cm^2 after eight months and persisted thereafter[28]. Normal forefoot pressures are generally less than 21 N/cm^2 [34]. Lengthening of tendo Achilles appeared to promote healing of forefoot ulcers in 80% of the patients treated[27], but resulted in substantial ankle weakness marked by reductions in the generation of plantar flexor torque during gait that persisted beyond eight months post-operatively[28], although by this time the forefoot pressure had returned to very high levels[28]. Other authors[27] found that 10% (seven of 73 ankles) suffered a rupture of tendo Achilles post-operatively, and 28% had dorsiflexion of > 15°, indicating a calcaneus gait and risking excessive heel pressures; 15% had a transfer of ulcers from the forefoot to the heel.

These complications may have devastating effects on walking, the recurrence of ulcers and on the future risk of amputation. Lengthening of tendo Achilles in diabetic individuals with equinus has some clear benefits, but also some risks. It is imperative that improved techniques be developed to optimise selection of cases so that only patients who would clearly benefit undergo the procedure.

The patients in previous studies were chosen for surgical intervention based on the presence of equinus as defined as < 5° of dorsiflexion on examination with a handheld goniometer. However, determination of equinus by physical examination of the range of movement on the ankle has some error[35]. The amount of dorsiflexion achieved is dependent upon the amount of force applied to dorsiflex the ankle, and determining the end of the range is subjective. Therefore, because of the large error associated with examination for equinus, some patients with equinus may have been inadvertently excluded from treatment, while others without equinus may have been unintentionally included. This lack of precision may lead to poor outcomes. My team developed an objective tool for measuring equinus (the Equinometer) that might aid in correctly identifying patients who would benefit from appropriate lengthening of the musculature of the triceps surae[30, 35-39]. This device aids the examining clinician to objectively apply a known load to dorsiflex the ankle for a specific length of time, and measure the subsequent range of the ankle with unprecedented accuracy.

The relationship between the degree of equinus present and the level of forefoot pressure during walking in diabetic individuals had not been rigorously established even though the connection was so well accepted that surgical lengthening to reduce forefoot pressure is an accepted practice in individuals with diabetes. My aim was to determine if there was a difference in peak forefoot pressure between patients with equinus and those without, as measured by the equinometer, and to use simple linear regression to establish the relationship between equinus and the level of peak forefoot pressure during walking in a group of patients with diabetes. The resulting paper was published in the Journal of Bone and Joint Surgery, British volume[40] and has been cited by 26 peer-reviewed publications. The data were presented at the American Orthopaedic Foot and Ankle Society Meeting and won the Mann Award for best clinical paper (Bruce Sangeorzan, MD, Senior AOFAS member).

The results of this study show that the degree of equinus (equinometer) does not have a substantial influence upon the level of forefoot plantar pressure during walking (Emed pressure mat; Novel, Munich, Germany). All participants in the study had forefoot pressures that were 2-3 times higher than normal values, suggesting an elevated ulceration risk. When grouped, those with equinus ($n = 13$; $2.1° \pm 2.0°$ ankle dorsiflexion at 10 Nm of dorsiflexing torque for 5 seconds) had increased forefoot pressures compared to those without equinus ($n = 14$; $8.1° \pm 2.9°$ dorsiflexion with 10 Nm torque). The pressure values were 63.9 ± 13.3 N/cm^2 for those with equinus vs. 50.9 ± 14.1 N/cm^2, for those without equinus; $p = 0.0245$. However, the degree of equinus predicted only 15% of the variance in forefoot pressure values and both the equinus and no equinus groups had forefoot pressure well above the normal value of 21Nm/cm^2.

The data from my study suggest caution in undertaking tendon-lengthening procedures until clearer indications are established. Judicious selection of individuals with equinus using an objective method is recommended before surgical interventions are undertaken to reduce forefoot pressure in diabetic subjects. Other factors like plantar tissue thickness[41] and mechanical characteristics[42, 43], bony deformity[44-47] and insensate plantar surface[48-51] may influence the formation of plantar ulcers.

Section 4. Quantifying the characteristics of functional locomotion

Orendurff MS, Schoen JA, Bernatz GC, Segal AD, Klute GK. How humans walk: bout duration, steps per bout, and rest duration. *J Rehabil Res Dev.* 2008;45(7):1077-1089.

Functional locomotion likely involves more than walking straight ahead at constant speed, the type of gait typically analyzed in a gait laboratory. Anyone who has observed, for instance, children with cerebral palsy in the gait analysis laboratory knows that their ability to complete the 180° turn at the end of the capture volume is a better differentiator of gait pathology than walking straight ahead: the most involved children take the longest and have the greatest difficulty during the turn. Once near-constant velocity has been achieved, walking straight ahead through the capture volume may not be as big a challenge as turning to children with cerebral palsy.

I hypothesized that most of human walking involves gait initiation from a standing or seated posture, accelerating to a desired speed, manoeuvring around objects and turning corners, decelerating, gait termination and returning to a standing or seated posture. This is the gait of daily living: the typical challenges that individuals must accomplish to locomotor transport in any environment. I assert that these tasks are considerably more complex and difficult to accomplish safely and efficiently with a neuromuscular pathology compared to walking straight ahead at self-selected speed. Altering the speed and trajectory of the body requires a wider array of joint motions, moments and powers to competently accomplish the desired change in location using bipedal gait.

There is very little published data that describes what human do, or need to do in order to be functional in their community and reach all the destinations and locations they desire during their daily activities. The community mobility challenge that most developed urban or suburban environments present defines the requirement for individuals to be fully mobile within such environments: the ecological validity of gait. Therefore, I designed a project to collect real-world locomotor data on a range of individuals without known pathology, but that represented a range of activity levels from highly active to highly inactive. My goal was to define how humans string steps together in order to move about during their day. If these gait parameters were accurately described there would be a specific metric of locomotor demand for developed urban and suburban settings for employed adults. This could be used as a design goal for interventions, and criteria for intervention success, along with technical metrics such as joint kinematics and kinetics from a gait laboratory and patient-reported outcomes. Although I had performed several different experiments on the effect of pathology on straight ahead walking, I needed to characterize the community mobility demand that each individual typically encounters to understand the effect of interventions on everyday ambulation.

I used StepWatch Activity Monitors to record the number of steps in each 10-second interval for a group of normal adults over a period of 14 days. I collected gait data until I had 10 weekdays and 4 weekend days, but not necessarily 14 consecutive days. Some individuals forgot to wear the StepWatch from time to time, and often it took three weeks of wear to acquire 14 days of data. The participants covered a range of activity levels, with some walking for exercise many times each week and sometimes several times per day, some activity in organized recreational sport competition and training, some active but non-sport participating, and some very inactive. The recorded data was processed in MATLAB to link together each sequential 10-second interval with steps, count the number of steps and the duration, in seconds. Output data (number of steps in each bout and the duration of the bout in seconds) was categorized using a frequency histogram of bout durations in thirteen, quasi-log scaled intervals.

Regardless of activity level, the results of this study show that human gait in typical western urban and suburban environments is dominated by very short walking bouts. A "walking bout" in this case is defined as a series of steps in sequence, with a beginning (gait initiation) and end (gait termination). The most frequent walking bout was 4±1 steps in a row, and occurs in about 15% of all walking bouts[52]. Walking bouts of 12 steps in a row or less account for 40% of all walking bouts in normal adults, and walking bouts of 40 steps in a row or less account for 75% of all walking bouts. A bout of 2000 steps in a row (about 15 minutes without stopping) occurs only once or twice per day for most people, and accounts for less than one half of one percent of all walking bouts. This gamma distribution of walking bouts length was observed in the most active individual who walked for health and sometimes performed 20,000 steps per day and in the least active individual who performed no exercise and less than 5000 steps on some days. Clearly, short durations of walking dominate human community ambulation for working adults in developed urban environments.

The data also reveal that 90.5% of all walking bouts are 4-99 steps in length, but this accounted for only 37.0% of total *steps*. Bouts of 100-199 steps in a row was 5.6% of total bouts, but accounted for 14.2% of all steps. Occasional long walking bouts

(1000-10,000 steps in a row) were extremely rare events (less than 0.6%), but accounted for 26.0% of all steps. The relationship between bout length and number of steps in that bout could form a precise global description of gait function within the community.

This paper was published in the Journal of Rehabilitation Research and Development and has been cited by 18 peer-reviewed publications covering wearable sensor validation, fall risk, amputee gait, arthroplasty interventions, stroke rehabilitation and spinal cord injury. In addition, my colleagues and I have developed several new and more specific methods for clearly describing a steps x bout duration x step rate "signature" for normal individuals, and for those with orthopaedic pathology. By including a method for incorporating step rate during the bouts as a measure of intensity a complete picture of all locomotor activity can be accurately quantified.

The data published in my paper offers some support to the hypothesis that the gait of daily living involves predominantly short bouts with gait initiation, turning, speed change, obstacle negotiation and gait termination highly in demand in typical urban settings for working adults. Additional work is needed to define the effectiveness of specific interventions designed to improve performance in specific components of this type of gait, and if this enables more competent ambulators following intervention. Although it is likely that three-dimensional computerized gait analyses will remain a mainstay of surgical decision-making for gait pathology, it appears prudent to link outcomes of technical measures such as kinematic or kinetic variables to performance on functional tasks that enable competent household and community ambulation, and improved patient satisfaction following interventions. Additional work on this topic using video analysis of real world settings was completed and published by Brian Glaister, a colleague and collaborator, in Gait and Posture[53]. This paper shows that for typical activities like purchasing items in a small shop, or eating in a cafeteria-style restaurant, short bouts of walking with frequent stops are required.

Section 5. Walking speed may influence plantar pressure

Segal A, Rohr E, Orendurff M, Shofer J, O'Brien M, Sangeorzan B. The effect of walking speed on peak plantar pressure. *Foot Ankle Int.* 2004;25(12):926-933.

With this data in hand, the next obvious challenge is to quantify the changes that occur in gait at faster walking speeds, then to investigate how individual might manipulate their ankle kinetics to alter their speed, and finally to investigate and describe how individuals utilize their ankle to turn and negotiate corners during the gait of daily living.

Walking speed may influence the pressure on the plantar surface of the foot, and may load some regions of the foot differently at faster walking speeds. The published data on walking at faster speeds suggests that the ankle moment is relatively constant, but that ankle power increases at faster walking speeds[54]. The increased ankle power at faster walking speeds may result in higher pressure on the plantar surface of the foot, or the constancy of the ankle moment at faster walking speeds may result in no change in plantar pressures at all. Therefore I initiated a study to quantify the effect of walking speed upon plantar pressure in functional regions of the foot. This paper was published in Foot and Ankle International[34] and has been cited in 29 subsequent peer-reviewed publications (see Appendix A). The data from this publication suggest that pressures under the plantar region of the heel and great toe are the most likely to be increased when maintaining faster walking speeds. This may be related to increases in stride length associated with faster walking speeds. However, the plantar surface of the metatarsal heads does not appear to be a region that bears increasing pressure at faster walking speeds. This has substantial impact for the formation of forefoot ulcers, and suggests that efforts to get individuals with diabetes to walk more slowly to reduce the risk of ulceration may be effective at the plantar surface of the heel and great toe, but may not to be as effective for reducing ulceration risk on the plantar surface of the metatarsal heads.

These data also suggest that the flexor hallicus longus and flexor hallicus brevus become proportionately more active and produce more plantar pressure under the great toe as faster walking speeds are maintained. This tends to increase the effective length of the foot, possibly to respond to the longer stride length present maintaining faster walking speeds. These data suggest that the first ray is the *performance ray* of the foot, becoming more active and functionally longer at faster walking speeds.

The citing papers cover a broad range of topics, but are primarily related to clinical issues in foot deformity[55], arthritis[56, 57], ontogeny[58-61], injury mechanisms and prevention[62, 63], and treatment outcomes[64]. One broad interpretation taken from my published paper is that studies of plantar pressure should include walking speed as a covariate for some regions of the plantar surface. The data from my study suggests that plantar pressures change at specific regions at different walking speeds, and this is related to the support and propulsion of the mass of the body during walking.

Section 6. Walking speed influences centre of mass motion

Orendurff MS, Segal AD, Klute GK, Berge JS, Rohr ES, Kadel NJ. The effect of walking speed on center of mass displacement. *J Rehabil Res Dev.* 2004;41(6):829-834.

In normal gait, the foot-ankle mechanism is principally responsible for the smooth motion of the centre of mass (COM) of the body, and for controlling forward progress at different walking speeds [4, 6, 7, 65-68]. If the ankle is dysfunctional in some way, other joints, primarily the hip, can assume some of the propulsive role of the ankle, but this appears much more energetically costly[69-76] and perhaps more complex from a controls standpoint. The movement of the COM had been accepted to be constant, moving about 2 cm vertically and mediolaterally during walking[77, 78] despite little published research to support this conclusion. This description has been included in at least thirteen textbooks on gait[78-90] even though data documenting the movement of the COM during walking had not ever been published in peer-reviewed literature.

Since stride length increases as walking speed increases, the position of the COM may achieve a lower position during double support at faster walking speeds compared with slower walking speeds. This suggests that the displacement of the COM might be greater at faster walking speeds. The movement of the centre of mass is supposed to be related to the energy efficiency of human walking, and the disruption of the inverted pendular motion may be an energetically costly consequence of ankle equinus, calcaneus gait (excessive dorsiflexion in late stance) or other ankle pathology.

During human walking the COM translates along the direction of travel but also moves in a sinusoidal pattern in the vertical and mediolateral directions. In both the vertical and mediolateral directions two maxima appear: The first near 30% of the gait cycle in single limb stance and another near 70% of the gait cycle in mid-swing; minima appear at 0% and 100% of the gait cycle in loading. Therefore, the COM reaches its highest and most lateral point as it passes over the planted foot and its lowest and most central point passing between one foot and the other. Since human walking is staggeringly complex, it has been suggested that analyzing the COM movement might simplify and illuminate the disturbances due to a broad range of pathologies [77].

The most efficient movement would theoretically result in forward translation of the COM without any vertical or mediolateral COM displacement. However, because of human anatomical structure and articulation geometry some vertical and mediolateral COM displacement is necessary to achieve forward progression. These mechanical constraints are integrated with control strategies to produce terrestrial locomotion that is economical in a wide array of bipedal, quadrapedal and polypedal organisms [91].

Despite the accepted usefulness of analyzing COM excursion to understand the overall impact on functional walking capacity due to movement pathologies, little is known about the effects of walking velocity on COM displacement. The ankle appears to influence the displacement of the COM, providing both support and propulsion that can be compromised in pathological gait. However, nearly all gait pathologies result in gait that is slower than normal. Differentiating between speed-related COM changes and pathology-related COM changes might be valuable for making treatment choices and understanding treatment outcomes. The purpose of this study was to examine the effect of walking speed on the COM displacement in the mediolateral and vertical directions. It was hypothesized that the COM would have equal vertical and mediolateral excursions at all walking speeds and that both would increase as walking speed increased.

The results of my study showed that the COM vertical excursion is smallest at slow walking speeds and increases with walking speed. The mediolateral excursion of the COM is greatest at slow walking speeds and decreases with faster walking speeds. Viewed from behind the COM forms a “U” shaped trajectory at faster walking speeds, but at slower speeds the minima begin to separate with the vertical minima occurring prior to the halfway point of the lateral excursion: at slow speeds the centre of mass drops more rapidly and then slowly climbs as weight is transferred to the contralateral limb. This publication has more clearly defined the motion of the COM during walking and the motion is not constant, but changes dramatically at different speeds.

This paper has been cited 28 times since its publication in the Journal of Rehabilitation Research and Development[92] (See Appendix A). The description of the movement of the COM during walking presented in my paper has now entered the gait lexicon as a “figure-of-eight” or “bow-tie” trajectory[93]. This has contributed to the understanding of normal gait function, and the concepts of COM in my paper have been used to gain better understanding of gait pathologies such as cerebral palsy[94] , fall risk in obese young adult males[95], and walking ability on challenging surfaces for those with lower extremity limb loss[96, 97]. Although I am pleased that several papers have cited my work, I will be most pleased with the normal path of the COM during gait in acknowledged as changing with walking speed in several of the monographs on gait mentioned previously.

In addition to gait biomechanics research, some authors have utilized the data in my paper to understand why wearable sensors that estimate activity levels using waist-worn 3D accelerometers are not accurate at slow speeds[98]. This error in estimating activity levels during low speed walking has been demonstrated by earlier work[99], but the errors were thought to be related to sensor technology. My paper demonstrates that mediolateral COM motion increases at slow walking speeds, but the vertical displacement diminishes; the accelerometers work as intended, but the motion of the waist does move in the same characteristic manner as in faster walking. With this understanding, alternate algorithms of 3D acceleration might lead to better assessments of activity levels at slow walking speeds with waist-worn accelerometers. My published work on COM motion through a range of walking speeds has penetrated a wide array of disciplines.

Section 7. Acceleration and deceleration while walking

Orendurff MS, Bernatz GC, Schoen JA, Klute GK. Kinetic mechanisms to alter walking speed. *Gait Posture.* 2008;27(4):603-610.

The next task was to address how individuals achieve the range of walking speeds used to in the gait of daily living. It was extremely surprising to me that there had been no study of how humans alter their speed while walking. I have search extensively for more than a decade to find such work with no success. The literature lacked even a basic hypothesis of how humans altered their walking speed. There were several publications examining the changes in joint motion, moments and powers while walking at different steady-state walking speeds[54, 100, 101], and several examining gait initiation[102-112] and gait termination[113-115], but no systematic investigations that even suggested a mechanism that humans typically used to increase or decrease their speed while walking. In earlier work I published, about 58% of all steps were within 2 steps of gait initiation or gait termination, suggesting that accelerating and decelerating are common tasks, more common than steps at constant speed. For household ambulation, about every fifth step is a turn[116], and turns require deceleration and acceleration. Based on these data it appears that altering walking speed might be a common task in everyday ambulation.

Perhaps due to our earliest work in gait, and due to the simple mechanical modelling of gait as an inverted pendulum that works out nicely from a mathematical standpoint, there is a tendency for investigators to concentrate on walking as a cyclic, patterned set of movements at constant speed in one direction – straight ahead. Almost no one has studied ecologically valid gait, which includes initiation, speed modulation, turning and gait termination starting from and returning to standing of seated postures.

Instead, most gait research has concentrated on straight ahead walking, and neglected the messy, chaotic and complex locomotor tasks of everyday ambulation. There is almost a Newtonian aspect to the thinking about gait, as if all acceleration occurs at gait initiation and the rest is coasting along effortlessly, never mind the string of braking and propulsive impulses during each stance phase while walking. Most computerized gait analysis laboratories do not have calibrated collection volumes that include the spaces at the beginning and end of their walkway. Individuals enter the capture volume at near constant speed and exit the collection volume before any substantial slowing. Studies of gait initiation and gait termination are published with much less frequency than studies of straight-ahead near-constant speed gait studies. Walking straight ahead at self-selected speed does not adequately differentiate between those with high and low functional abilities, and may not fully describe the performance during the challenges typically encountered during daily movement within the community.

The scientific community has recognized that slow gait is not functional and that individuals who cannot maintain functional walking speeds have greater disability than healthy adults and children, but has not studied the (dysfunctional) mechanisms individuals use to modulate their speed. Although several key papers have described the kinematic and kinetic change that occur as individuals *maintain* faster walking speeds in pathological, elderly and normal individuals, *changes* in speed while walking have been unexplored. Other than for initiation and termination of gait, *how* individuals accelerate and decelerate to achieve functional walking speeds has not been investigated in the published peer-reviewed scientific literature, even though it clearly may improve our understanding of how pathology affects mobility.

Therefore, I undertook a project to describe how human accelerate and decelerate during walking. The experiment involved walking through a 3D motion capture volume with four distinct profiles: 1.0 m/s at constant speed; 1.4 m/s at constant speed; acceleration from 1.0 to 1.4 m/s; and deceleration from 1.4 to 1.0 m/s. For each condition, participants (n = 12) contacted the force plates, and only trials that had the appropriate speed change profile were retained for analysis. Trials with "targeting" by observation or participant report were discarded. All acceleration or deceleration trials had foot contact on the force plates before any speed change occurred, to ensure that the entire load path was measured during speed change. A full body marker set was utilized and COM location was calculated for each participant using segmental analysis and the segment parameter data of Dempster.[117]

It was hypothesized that walking acceleration would occur by the individual creating greater push-off power with their ankle late in stance phase, as is the case for faster walking compared with slower walking[54]. This would result in greater propulsive impulses (area under the anterior-posterior force-time curve) and greater peak sagittal ankle power values in late stance. Deceleration was hypothesized to be achieved by the opposite effect: less ankle push-off during late stance phase. It was hypothesized that peak ankle power in late stance would be greatest in acceleration, lower in walking fast at constant speed, lower still during deceleration and lowest during slow walking.

However, after examining the statistical results and interpreting the data, it was obvious that the change in walking speed during acceleration and deceleration trials occurred well before peak ankle push-off power was observed. The onset of speed change occurred in single limb stance and corresponded to a significant but somewhat subtle change in the ankle moment pattern: the slope of the ankle moment was concave during acceleration trials and convex during deceleration trials. These differences in the sagittal ankle moment value at 30% of the gait cycle showed that fast and slow walking had similar values, that acceleration trials had decrease values only on the second stride when acceleration began, and that deceleration trials were lower than fast or slow walking on both the first and second stride (deceleration occurred over two steps).

This pattern is very much like that seen in mild equinus: an ankle moment curve with a convex shape and a steeper rise rate in single limb stance generally causes a slowing of the body during walking. Therefore, decelerating gait speed while walking appears to be accomplished by making subtle changes in the force of the plantarflexors and the ankle moment generated in midstance. Acceleration is the opposite of this: reduced plantarflexor force and reduced plantarflexor moment in single limb stance. This alters the anterior-posterior ground reaction (braking) impulse imparted to the body during early stance; deceleration increases and prolongs the braking impulse in early stance, and acceleration reduces and shortens the braking impulse. Acceleration from 1.0 m/s to 1.4 m/s can occur in a single step, but deceleration appeared spread over two steps for 1.4 m/s to 1.0 m/s change.

These data also challenge a long held theory of human gait: controlled falling. This theory posits that humans utilize gravity to accelerate the centre of mass downward, arcing on their stance limb as if they were falling, and then catch themselves with their contralateral (lead) limb, transferring momentum from the vertical direction to the forward direction. It does seem an attractive mechanism for acceleration during walking: delay lead-limb foot contact, and have the COM accelerated downward by gravity for a longer period of time to increase walking speed, but this was not observed in the data collected. There were no statistically significant differences observed in the position of the COM at 30% (highest point) and 50% (lowest point) of the gait cycle during slow, fast, acceleration or deceleration walking. It does not appear that normal humans use this mechanism to increase or decrease walking speed, or to walk at faster or slower speeds over the range of speeds in this study. It is more likely that controlled falling is used for gait initiation. My previous work has shown that the COM always reaches a similar maximum height during single limb stance with full knee extension, but that the minimum becomes lower and lower at faster and faster walking speed, due primarily to increasing step length. This is not to say that humans do not perceive and take advantage of the characteristics of the acceleration due to gravity to ambulate, but we do so in a much more complicated manner than originally hypothesized.

Based on the data from my study it is more correct to describe human gait as controlled rising, since this is how humans appear to modulate their speed. Subtle but significant manipulation of the braking impulse in early stance by the ankle plantarflexors either increases (deceleration) or decreases (acceleration) the braking impulse while the centre of mass is moving from its lowest position during double support to its highest position in single limb stance. This is accomplished by altering the forces and moments at the ankle joint, but not by altering the motions of the ankle joint: the sagittal motion of the ankle in all conditions was not significantly different. It is unlikely that increased plantarflexion push-off is used as a mechanism to accelerate or decelerate because by the time late stance occurred in the acceleration or deceleration trials, the speed change had already taken place.

Acceleration from 1.0 m/s to 1.4 m/s was accomplished in a single stride, but deceleration was spread over two strides. The ankle moment in single limb stance was not different during slow or fast walking, only peak ankle power generation was different between fast and slow. Earlier work by Lelas et al has shown that peak ankle power increases with walking speed[54], and taken together these data suggests that manipulating the ankle moment in single limb stance is the most likely mechanism of modulating speed while walking, and that increased ankle power generation in late stance accommodates and maintains walking speed.

At first glance this seems highly complex from a controls standpoint. This suggests that humans may have one mechanism to change walking speed and a different mechanism to accommodate to current walking speed. But in human gait, this separation of the two might be more economical from both an energetic-metabolic cost standpoint and a control standpoint. The data from my study of acceleration and deceleration suggests that by removing a force (perhaps doing less mechanical work) leads to acceleration while walking. This strategy also applies forces evenly over a longer period, perhaps making smooth acceleration more likely and easier to plan adaptive strategies for possible balance perturbations or execution errors.

This proposed strategy also makes changing walking speed similar at all different walking speeds. Rather than relying on increasing ankle push-off power, which already has to cope with faster walking speeds, the change of walking speed is performed by a different kinetic mechanism. This allows the ankle to alter speed with one mechanism, regardless of current speed, and allows the ankle to cope with current walking speed using a different mechanism. These are also separated in time, with speed changes occurring in single limb stance, and adapting to current speed occurring in late stance. Both of these mechanisms are controlled by the ankle plantarflexors.

This paper was published in Gait and Posture and has been cited by two subsequent publications to date. In contrast to my findings, the recent work by Peterson, Kautz and Neptune[118] found somewhat different results using a force plate-equipped treadmill to study acceleration and deceleration. The sagittal ankle moment did not appear to adapt in the same way to increasing walking speed, and both acceleration and deceleration moments had a concave appearance. The reason for this may be due to the power imparted to the individual by the mass of the treadmill. Riley, et al have shown that the sagittal ankle moment has similar peaks in early and late stance in treadmill and overground walking, but the appearance of the single limb stance moment pattern is different[119]. It appears that on the treadmill, the normal human response is to attenuate the excessive acceleration transferred to the individual by the treadmill during single limb stance. Treadmill gait also has much lower step-to-step variation in joint kinematics and kinetics. Riley's intent was to show that treadmill walking was identical to overground walking, however these authors only compared minimum or maximum moments, and did not subject other portions of the curves to hypothesis testing.

Another reason that Peterson, Kautz and Neptune[118] found no change in the ankle moment pattern was the rate of acceleration was very slow, 0.12 m/s^2 and this is so subtle as to mask the mechanism. In my study, the speed change was about 0.2 to 0.4 m/s^2, and this may account for the differences between the results of the two studies. Perhaps this is due to the use of a treadmill, since the participants are adapting to a slow external acceleration whereas in my study the participants were initiating their speed change themselves. Perhaps these two mechanisms operate in different situations, with larger speed changes controlled by the ankle moment in single limb stance and smaller speed changes adapted to in different but as yet poorly defined ways.

Section 8. Turning while walking

Orendurff MS, Segal AD, Berge JS, Flick KC, Spanier D, Klute GK. The kinematics and kinetics of turning: limb asymmetries associated with walking a circular path. *Gait Posture.* 2006;23(1):106-111.

Based on my previous work in the gait of daily living[52], and the work of others[116], turning seemed to be a task that was frequently encountered in household and community ambulation, was challenging, but very little was known about how humans turn.

For example, falls in the elderly are much more likely to occur while turning and the injuries are more severe[120]. The extent of functional limitation may be more observable when an individual is performing an activity other than walking straight ahead. Wall et al have shown that during a timed get-up-and-go test, the turn differentiated older, frail adults better than any other component of the task[121]. The possibility that the short duration manoeuvring movements might be more challenging suggests that gait therapy might be more effective if focused on these more challenging movements to improve functional mobility in those with gait pathology. This proposition is supported by the fact that most living spaces do not have long straight areas where straight ahead walking dominates the repertoire of movements needed, but rather our homes, places of business and healthcare, shops, stores and communities have various barriers that require us to turn and manoeuvre to the get to our desired location. Improving turning ability may improve an individual's functional gait performance.

Focusing on competent turning and manoeuvring during gait therapy is difficult because the characteristics of the task have not yet been defined. For example strengthening a specific joint, sit-to-stand practice, standing balance training, gait initiation or gait termination practice, or some optimal combinations of these might be more effective or advantageous than just working on walking straight ahead. These therapeutic choices are impossible to make without some evidence of the challenges of the task individuals will face in their everyday lives. However, little was known about the biomechanics of how humans turn, how they alter their limb kinematics and kinetics to alter the path of the centre of mass, and how they rotate their body during turning to face the direction of travel.

Therefore, I initiated an experiment to describe how humans turn. The experiment tested four hypotheses: turning was accomplished by altering lateral forces by more lateral foot placement; turning was accomplished by increased push-off power on the outside ankle; turning was accomplished by functional shortening of the inside limb; turning was accomplished by transverse plane torques that rotated the trunk. The most substantial of these with the greatest F-test value was the lateral shear impulses that act to shift the body in the direction of the desired turn. This suggests foot placement in the medial-lateral direction during walking was a key component of turning gait, with other mechanisms more likely to be a response to foot placement: the inside limb did functionally shorten (more joint flexion) and the outside limb lengthened (more joint extension), but these effects appeared less dramatic, although also statistically significant. The transverse plane hip rotator torques appeared to play a role in the orientation of the trunk, as suggested from this manuscript, and in initiating and terminating the rotation of the body in a 90° hallway corner[122, 123]. The paper was published in Gait & Posture[11] and has been cited by 30 subsequent publications (See Appendix A).

The papers that cite my paper on turning included experimental work on turning on individuals with Parkinson's[124, 125], robotic walking principals[126-128], total joint design and function during activities of daily living[129], individuals recovering from a stroke[130], and biomemetic prosthetic ankle design[131]. Several other investigators at different laboratories have initiated robust research programs investigating human turning and this indicates a novel research field in its own right. Several of my collaborators have also continued this line of research. Brian Glaister has published several interesting papers on turning, creating an entire career on this area. His work defines the ground reaction forces and impulses during a hallway turn[122], quantifying the transverse plane mechanical behaviour of the human ankle joint using elastic and viscous behaviour modelling within 1-4 states in the initiation, apex and terminations steps, and a video task analysis of humans walking in ecologically valid community walking environments. This latter publication is another validation that turning is a frequent task needed in everyday ambulation. Glaister has also published a fascinating methods paper on how to rotate the ground reaction forces due to turning[132]; this keeps the lateral forces lateral and the fore-aft forces force-aft no matter how the individual turns while on a force plate.

Conclusion

At the beginning of this series of experiments my over-arching thesis was that the gait of daily living involves gait initiation, acceleration to the desired speed, deceleration—obstacle-negotiation/turning—acceleration, deceleration and gait termination from standing or seated postures. I hypothesized that these challenges were different and more complex than walking straight ahead at self-selected speed. I hypothesized that the ankle was central to both simple and more complex locomotor tasks. To provide support for these hypotheses, I began with normal plantarflexor function during walking at constant speed, investigated pathologic populations with equinus during walking at constant speed, acquired data that demonstrated the gait of daily living involved many short locomotor bouts with frequent short rests and many speed changes. Then I investigated the effect of maintaining different walking speeds on plantar pressures and COM motion, how humans might use their ankle to accelerate and decelerate, and finally how humans might turn. The data generated from these experiments has define the force-length-velocity characteristics of normal plantarflexors during walking, demonstrated that judicious tendo Achilles lengthenings may be appropriate for some children with equinus, and that objectively measured equinus in diabetic individuals may not necessarily lead to an increase in already elevated forefoot pressures. I have shown that locomotion in ecologically valid environments involves short bouts with frequent stops and starts. The data generated have also shown that speed influences plantar pressure on some regions of the foot, that speed alters the path of the COM, that the ankle moment may control acceleration and deceleration while walking, and offered a theory of how humans manipulate their joint kinematics and kinetics to alter their trajectory and orientation to avoid obstacles and negotiate turns.

This supporting statement covers a range of published investigations into the role of ankle function in complex types of human walking, in both normal individuals and those with pathology. My published work includes both laboratory-based technical measures of gait, and addresses novel human gait characteristics that appear to be used in everyday locomotor activities (see Appendix A). My work has stirred considerable technical and clinical discourse, and has been cited by 180 peer-reviewed publications to date. This evidence suggests that my work has had some impact on the field.

Appendix A: Published papers upon which this supporting statement is based

Orendurff MS, Segal AD, Aiona MD, Dorociak RD. Triceps surae force, length and velocity during walking. *Gait Posture.* 2005;21(2):157-163.

Yuen TJ, Orendurff MS. A comparison of gastrocnemius muscle-tendon unit length during gait using anatomic, cadaveric and MRI models. *Gait Posture.* 2006;23(1):112-117.

Orendurff MS, Aiona MD, Dorociak RD, Pierce RA. Length and force of the gastrocnemius and soleus during gait following tendo Achilles lengthenings in children with equinus. *Gait Posture.* 2002;15(2):130-135.

Orendurff MS, Rohr ES, Sangeorzan BJ, Weaver K, Czerniecki JM. An equinus deformity of the ankle accounts for only a small amount of the increased forefoot plantar pressure in patients with diabetes. *J Bone Joint Surg Br.* 2006;88(1):65-68.

Orendurff MS, Schoen JA, Bernatz GC, Segal AD, Klute GK. How humans walk: bout duration, steps per bout, and rest duration. *J Rehabil Res Dev.* 2008;45(7):1077-1089.

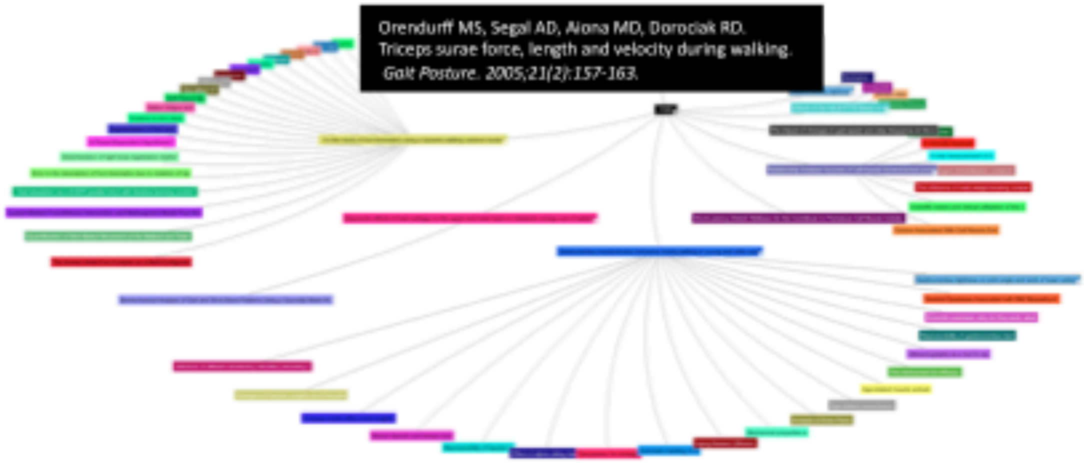
Segal A, Rohr E, Orendurff M, Shofer J, O'Brien M, Sangeorzan B. The effect of walking speed on peak plantar pressure. *Foot Ankle Int.* 2004;25(12):926-933.

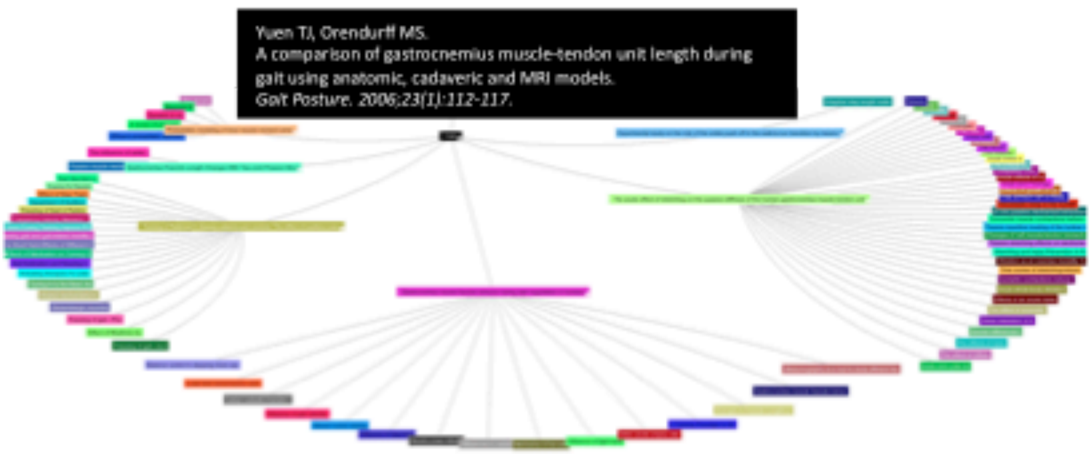
Orendurff MS, Segal AD, Klute GK, Berge JS, Rohr ES, Kadel NJ. The effect of walking speed on center of mass displacement. *J Rehabil Res Dev.* 2004;41(6):829-834.

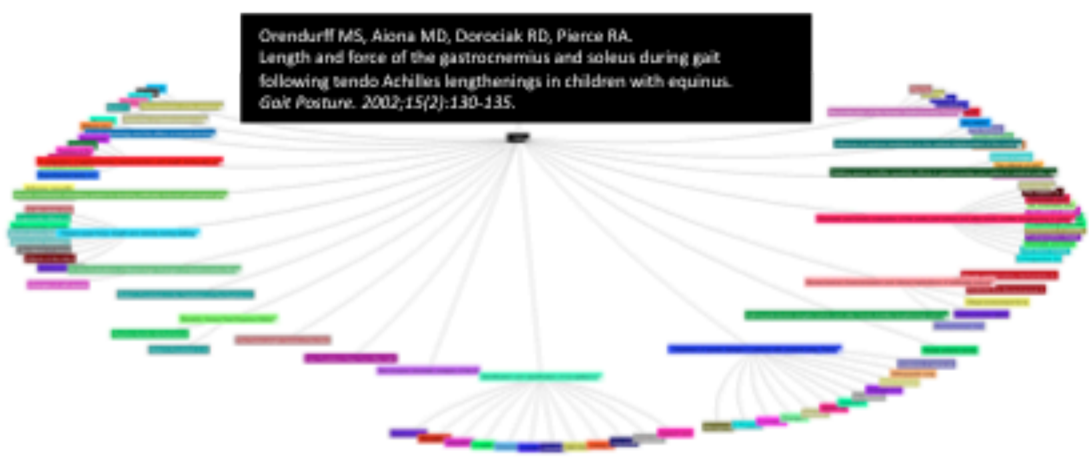
Orendurff MS, Bernatz GC, Schoen JA, Klute GK. Kinetic mechanisms to alter walking speed. *Gait Posture.* 2008;27(4):603-610.

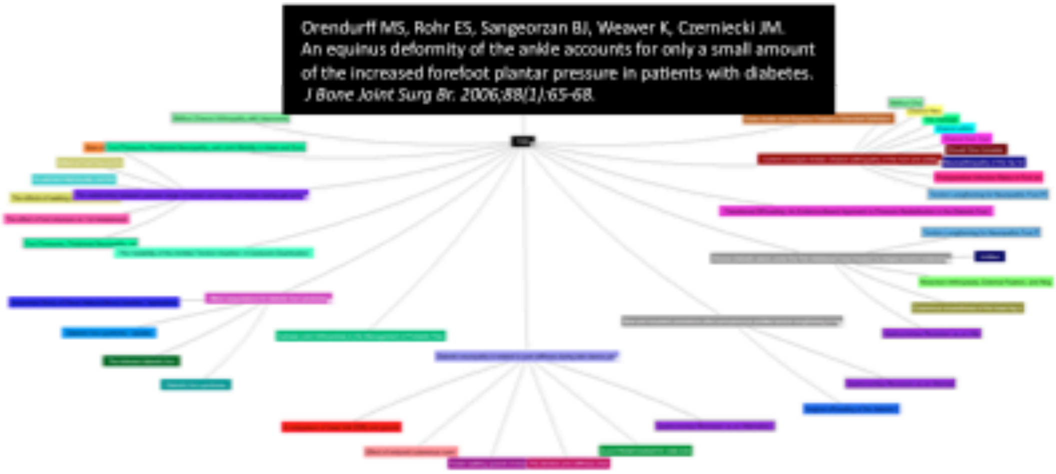
Orendurff MS, Segal AD, Berge JS, Flick KC, Spanier D, Klute GK. The kinematics and kinetics of turning: limb asymmetries associated with walking a circular path. *Gait Posture.* 2006;23(1):106-111.

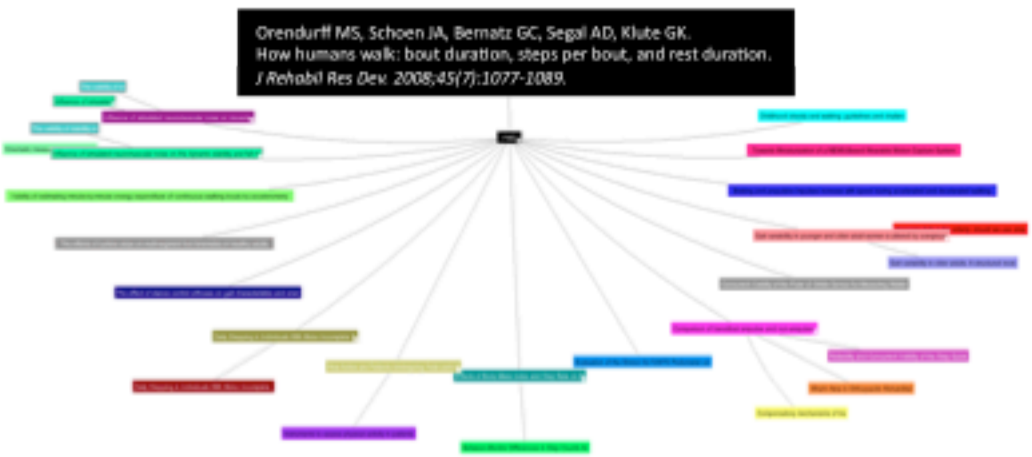
Graphical representations of my publications and the subsequent publications that have cited my work to date. Data from Science Citation Index.








Orendurff MS, Rohr ES, Sangeorzan BJ, Weaver K, Czerniecki JM.
An equinus deformity of the ankle accounts for only a small amount of the increased forefoot plantar pressure in patients with diabetes.
J Bone Joint Surg Br. 2006;88(1):65-68.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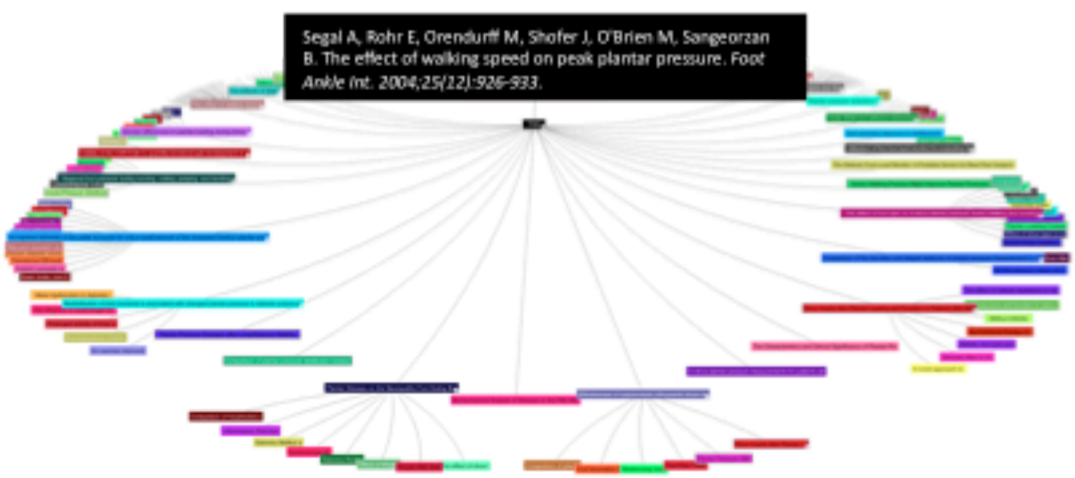
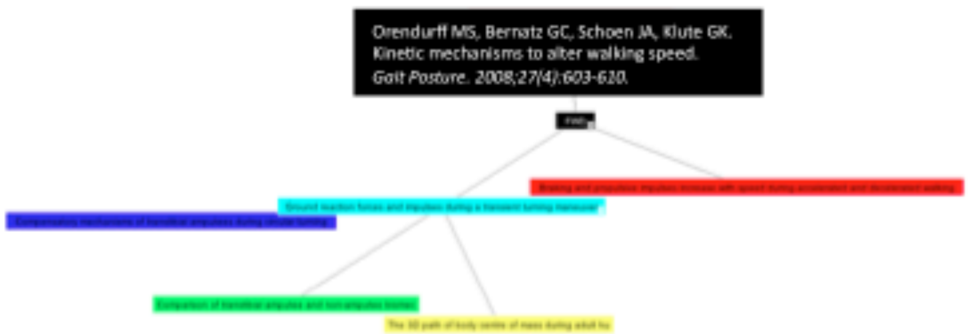
How humans walk: bout duration, steps per bout, and rest duration.
J Rehabil Res Dev. 2008;45(7):1077-1089.


Segal A, Rohr E, Orendurff M, Shofer J, O'Brien M, Sangeorzan B. The effect of walking speed on peak plantar pressure. Foot
Ankle Int. 2004;25(12):926-933.


Orendurff MS, Segal AD, Klute GK, Berge JS, Rohr ES, Kadel NJ.
The effect of walking speed on center of mass displacement.
J Rehabil Res Dev. 2004;41(6):829-834.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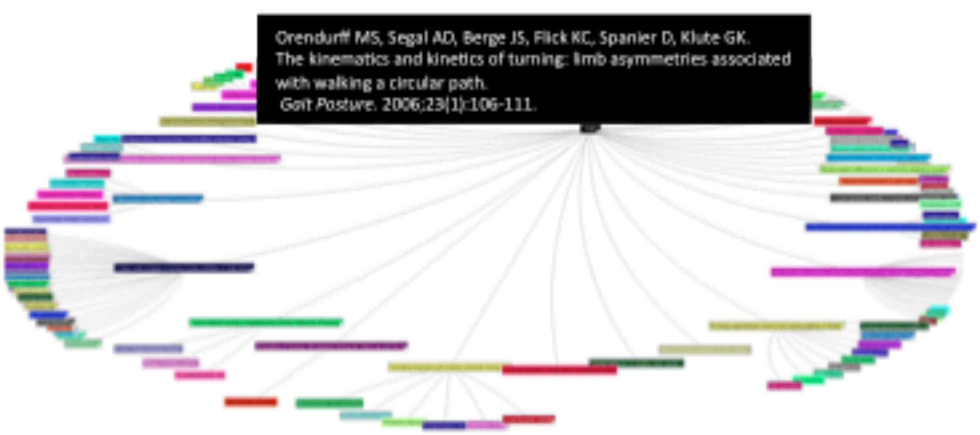
Kinetic mechanisms to alter walking speed.
Gait Posture. 2008;27(4):603-610.


Orendurff MS, Segal AD, Berge JS, Flick KC, Spanier D, Klute GK.
The kinematics and kinetics of turning: limb asymmetries associated with walking a circular path.
Gait Posture. 2006;23(1):106-111.

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